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High Specific Activity Chemiluminescent and Fluorescent Markers: their Potential Application to High Sensitivity and 'Multi-analyte' Immunoassays

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The sensitivities of immunoassays relying on conventional radioisotopic labels (i.e. radioimmunoassay (RIA) and immunoradiometric assay (IRMA)) permit the measurement of analyte concentrations above *ca* 10^7 molecules/ml. This limitation primarily derives, in the case of 'competitive' or 'limited reagent' assays, from the 'manipulation errors arising in the system combined with the physicochemical characteristics of the particular antibody used; however, in the case of 'non-competitive' systems, the specific activity of the label may play a more important constraining role. It is theoretically demonstrable that the development of assay techniques yielding detection limits significantly lower than 10^7 molecules/ml depends on:

(1) the adoption of 'non-competitive' assays designs;
(2) the use of labels of higher specific activity than radioisotopes;
(3) highly efficient discrimination between the products of the immunological reactions involved.

Chemiluminescent and fluorescent substances are capable of yielding higher specific activities than commonly used radioisotopes when used as direct reagent labels in this context, and both thus provide a basis for the development of 'ultra-sensitive', non-competitive, immunoassay methodologies. Enzymes catalysing chemiluminescent reactions or yielding fluorescent reaction products can likewise be used as labels yielding high effective specific activities and hence enhanced assay sensitivities.

A particular advantage of fluorescent labels (albeit one not necessarily confined to them) lies in the possibility they offer of revealing immunological reactions localized in 'microspots' distributed on an inert solid support. This opens the way to the development of an entirely new generation of 'ambient analyte' microspot immunoassays permitting the simultaneous measurement of tens or even hundreds of different analytes in the same small sample, using (for example) laser scanning techniques. Early experience suggests that microspot assays with sensitivities surpassing that of isotopically based methodologies can readily be developed.

Keywords: Ultrasensitive immunoassay; fluorescent microspot immunoassay; confocal microscopy

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INTRODUCTION

Immunoassay methods relying on radioisotopic labels have played a major role in medicine and other biologically related fields (agriculture, veterinary science, the food and pharmaceutical industries, etc.) during the past two decades. Their importance has derived from the exploitation both of the 'structural specificity' characterizing antibody–antigen reactions and the 'detectability' of isotopically-labelled reagents, the latter permitting observation of the binding reactions between exceedingly small concentrations of the key reactants involved. The combination of these features has endowed radioimmunoassay methods with unique specificity and sensitivity characteristics, and accounts for their ubiquitous use throughout modern medicine and biology. However, in the past few years, interest has increasingly focused on so-called 'alternative', non-radioisotopic, immunoassay methods; such techniques are based on essentially identical analytical principles but differ in the markers used to label the particular immunoreactant (antibody or analyte) whose distribution between bound and free moieties (following the basic analytical reaction) constitutes the assay 'response'. The reasons for this interest may be grouped under four headings:

(1) Environmental; logistic; economic; practicality and convenience, etc. (i.e. 'non-scientific).
(2) The attainment of higher sensitivity.
(3) The development of 'immunosensors' and 'immunoprobes'.
(4) The development of 'multi-analyte' assay systems.

Our own reasons for developing non-isotopic techniques fall principally under headings (2) and (4), and this presentation will centre primarily on the concepts which underlie our immunoassay development strategy in these areas.

THE ATTAINMENT OF 'ULTRA-HIGH' IMMUNOASSAY SENSITIVITY

Though, as indicated above, the sensitivity of radioisotopically based immunoassay methods has constituted one of the principal foundations of their widespread use over the past 25 years, a fundamental reason for their replacement stems, paradoxically, from the current requirement to develop microanalytical techniques which are superior to them in this particular respect. Radioisotopic methods are, in practice, limited to the measurement of analyte concentrations above about 10^8–10^9 molecules/ml (i.e. approx 0.15–1.5 pmol/l)(Dakubu *et al.*, 1984). However, in certain fields (e.g. virology, tumour detection) there is a particular need to detect or measure molecular concentrations below this level. The factors which determine immunoassay sensitivity have been extensively discussed (Ekins *et al.*, 1968, 1970a; Ekins, 1978; Jackson *et al.*, 1983; Dakubu *et al.*, 1984; Ekins, 1985). Nevertheless, some of the underlying concepts are still frequently misunderstood and merit brief discussion in the present context.

The concept of sensitivity

One major source of past confusion has been disagreement regarding the concept of 'sensitivity' itself, many authors equating assay sensitivity with the slope of the dose–response curve (Yalow and Berson, 1970a, b; Berson and Yalow, 1973; see also Ekins *et al.*, 1970b, Tait, 1970). It is now widely agreed that the notion that a steeper dose–response curve implies greater sensitivity is erroneous. The invalidity of this belief is clearly revealed by the fact that the relative magnitudes of the responses yielded by two assay systems is dependent on the particular variable which is chosen to represent the response (see Fig. 1(a))(Ekins, 1976). For this and other reasons, it has long been recognized that the 'sensitivity' of an assay can only be satisfactorily represented by its lower limit of detection (Fig. 1(b)), and this concept is now embodied in all internationally agreed definitions of the term. An essentially identical definition is as the precision (i.e. standard deviation) of measurement of zero dose, since this quantity determines the least quantity distinguishable from zero and hence the assay detection limit. The sensitivity of an assay is thus represented by the zero-dose intercept of the 'precision profile' (Fig. 2(a)) when the latter is expressed in terms of standard deviation rather than of coefficient of variation (Ekins, 1983a). In short, the more sensitive of two assays is the one yielding greater precision of the zero dose estimate (Fig. 2(b)).

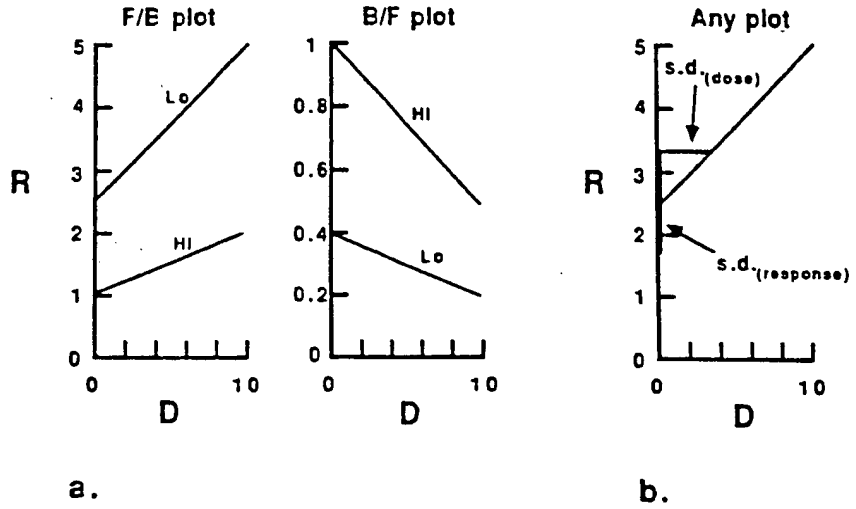

Figure 1. (a) Diagrammatic representation of conventional RIA dose–response curves for systems using high (hi) and low (lo) antibody concentrations plotted in terms of free-bound (F/B) and bound/free (B/F) labelled antigen. Note that the use of a lower amount of antibody yields a dose–response curve of greater slope in the F/B plot, but of lower slope in the B/F plot. It is impossible to decide, on the basis of the data shown in this figure, which concentration of antibody yields the assay system of higher sensitivity. (b) The sensitivity of an assay is essentially represented by the minimum detectable dose, i.e. the SD of the dose measurement ($SD_{(dose)}$) at zero dose. This is given by the SD of the response ($SD_{(response)}$) divided by the dose–response curve slope at zero dose (i.e. $(SD_{(response)}) \times dD/dR)_0$). This quantity is unaffected by the choice of the coordinate frame used to plot the dose–response curve. (Note: it is common to multiply $(SD_{(dose)})_0$ by an arbitrary factor to increase the confidence level attaching to the minimum detectable dose estimate, though, since no agreement exists regarding the value of this factor, this unnecessary step merely adds to confusion when the relative sensitivities of two assay procedures are compared.)

'Competitive' and 'non-competitive' ('limited reagent' and 'excess reagent') assays

A second important misconception in this area is the notion that immunoassays relying on the use of *labelled antibodies* (e.g. immunoradiometric assays, IRMA) are *ipso facto* more sensitive than those which rely on the use of *labelled 'analyte'* (e.g. radioimmunoassays, RIA); furthermore the grounds originally advanced for the claimed superiority of labelled antibody methods (Miles and Hales, 1968) were partially based on false concepts of sensitivity, and thus failed to identify the *true* reasons why certain assay designs are

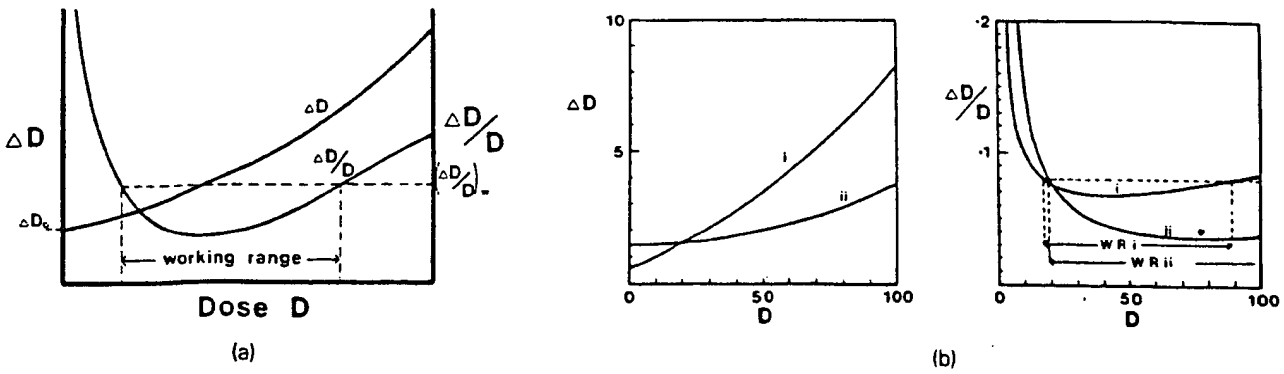

Figure 2. (a) The 'precision profile' of an assay portrays the error in the dose measurement as a function of dose. The error may be represented, *inter alia*, by the absolute error (ΔD; e.g. SD of D) or the relative error ($\Delta D/D$; e.g. CV of D). $(\Delta D)_0$, the error in the measurement of zero dose, represents the sensitivity of the assay. The working range may be defined as the range of dose values within which $\Delta D/D$ is less than an 'acceptable' value set by the investigator. (b) The more sensitive of the two assays (assay I) intercepts the ΔD axis at a lower value. However, assay II is more precise at higher values of dose, and has a wider working range.

potentially capable of yielding far higher sensitivity than others. This issue likewise merits clarification.

The purely pragmatic sub-classification of immunoassays into labelled antibody and labelled analyte methods diverts attention from a more fundamental divide in immunoassay methodology, which relates to the optimal concentration of antibody required in an assay system to maximize its sensitivity. In certain assay designs (which may be termed '*limited reagent*' or '*competitive*') the optimal concentration tends to zero; conversely in others (which may be termed '*excess reagent*' or '*non-competitive*') the concentration tends to infinity. It should be particularly emphasized that the optimal antibody concentration is essentially governed, not only by the physicochemical characteristics of the antibody–analyte binding reaction, but also by the errors incurred in measurement of the assay response. Were an assay system to be totally error-free, *no* antibody concentration would be optimal, and the distinction between competitive and non-competitive methodologies would thus not arise.

Though it is inappropriate in this presentation to discuss in detail the statistical and physicochemical theory underlying this fundamental divergence in immunoassay design (see Ekins *et al.*, 1968, 1970a; Jackson *et al.*, 1983), the reason for it can perhaps be more readily understood if the basic principles of immunoassay are portrayed in a somewhat different way from that in which they are usually presented. All immunoassays essentially depend upon measurement of the 'fractional occupancy' by analyte of antibody binding sites following reaction of analyte with antibody (see Fig. 3(a)). Those techniques which implicitly rely on measurement of residual, *unoccupied*, binding sites optimally necessitate the use of concentrations of antibody tending to zero, and may be termed 'competitive', conversely those in which *occupied* sites are directly measured necessitate use of high antibody concentrations and are termed 'non-competitive' (Fig. 3(b)). This emphasizes that the differences in assay design characterizing so-called competitive and non-competitive methods are essentially unrelated to which component (if any) of the reaction system is labelled. Indeed immunoassays in which *no label of any kind is involved* can, on identical grounds, be subdivided into those of 'limited reagent' (or 'competitive') and 'excess reagent' (or 'non-competititve') design. Thus the distinction between these two forms of immunoassay simply reflects differences in the way that fractional antibody occupancy is determined, and the fact that it is generally undesirable—for reasons of accuracy—to measure a *small* quantity by estimating the difference between two *large* quantities. When an immunoassay relies on the measurement of unoccupied antibody binding sites, the total amount of antibody used in the system must be small to minimize error in the resulting (indirect) estimate of occupied sites.

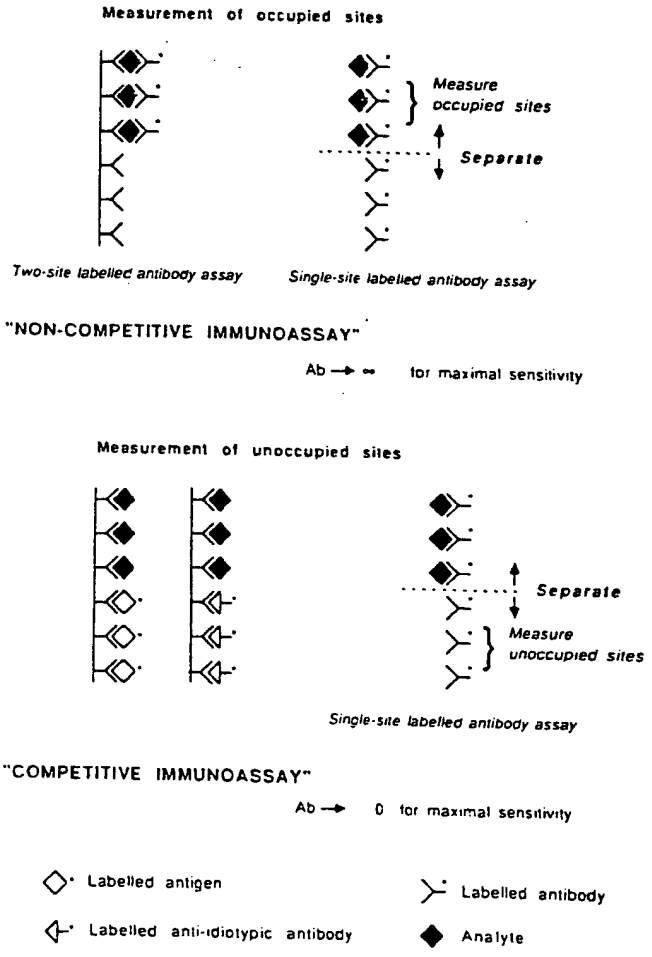

Figure 3. The distinction between 'non-competitive' (above) and 'competitive' immunoassays (below) reflects how antibody binding-site occupancy is measured. Labelled antibody methods are 'non-competitive' if occupied sites of the (labelled) antibody are measured, but are 'competitive' (below right) when *unoccupied* sites are measured. Labelled antigen (below left) or labelled anti-idiotypic antibody methods (below centre) rely on measurement of sites *unoccupied* by analyte, and are therefore invariably of 'competitive' design.

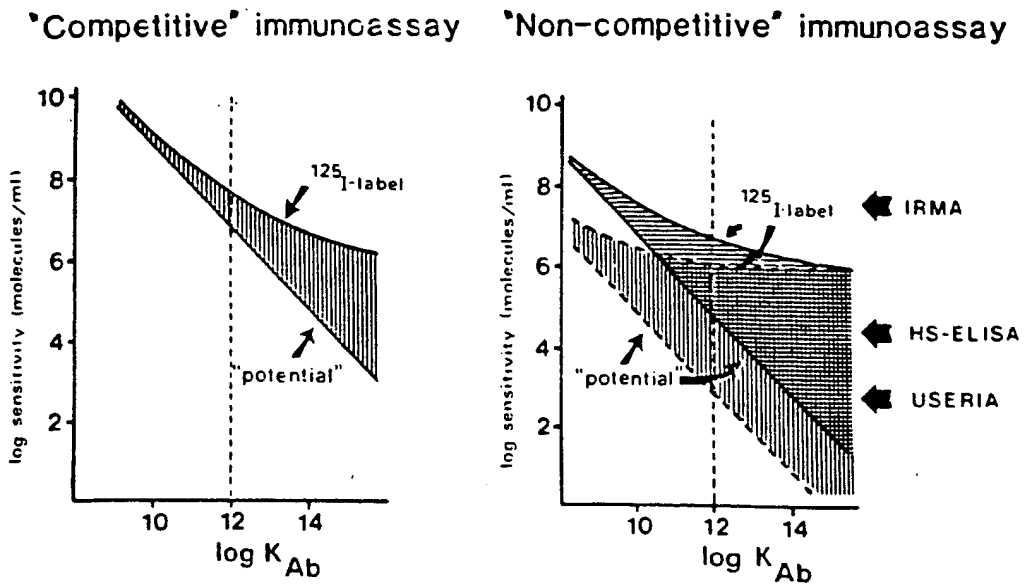

Figure 4. Curves showing the theoretically predicted relationship between antibody affinity and the sensitivities achievable using 'competitive' and 'non-competitive' assay strategies. The 'potential' sensitivity curves assume the use of infinite specific activity labels; the sensitivities achievable using ^{125}I-labelled antigen or antibody are also shown. Shaded areas indicate the sensitivity loss due to errors in measurement of the label. Curves relating to 'competitive' assays assume a 1% error in measurement of the response variable arising from 'experimental' errors (i.e. errors other than those inherent in label measurement *per se*). Non-competitive curves assume 'non-specific binding' of labelled antibody of 0.01% and 1% (lower and upper curves) respectively. Arrows indicate sensitivities claimed for typical non-competitive immunoassay methodologies.

Conversely, when occupied sites are measured *directly*, this particular constraint does not arise; indeed, considerable advantage often derives from using relatively large amounts of antibody in the system.

Sensitivity of 'competitive' and 'non-competitive' immunoassays

Competitive and non-competitive immunoassays differ significantly in many of their performance characteristics in consequence of the differences in optimal antibody concentration on which they rely. Most particularly they differ in their potential sensitivities. Figure 4. portrays the sensitivities predicted theoretically as a function of antibody binding affinity, making realistic assumptions regarding the experimental errors incurred in reagent manipulation, 'non-specific' binding of labelled antibody, etc., and assuming the use of optimal reagent concentrations (Ekins, 1985). Amongst other concepts illustrated in the figure is the much greater assay sensitivity *potentially* attainable (using an antibody of given affinity) by adoption of a non-competitive approach. In short, whereas the maximal sensitivity realistically achievable using a competitive design is in the order of 10^7 molecules/ml (using antibody of the highest affinity found in practice), a non-competitive method is capable of yielding sensitivities some orders of magnitude greater than this. However, Fig. 4 also demonstrates that, assuming the use of high affinity antibodies (i.e. $\sim 10^{11}$–10^{12} l/M), maximal sensitivities yielded by isotopically based techniques (whether relying on labelled antibody (IRMA) or labelled analyte (RIA), or whether of competitive or non-competitive design) are closely comparable, i.e. of the order of 10^7–10^8 molecules/ml.

This limitation is a manifestation of the fact that, in the case of the non-competitive methods, an important constraint on assay sensitivity is (under certain circumstances) the 'specific activity' of the label used. On the other hand, limitation of assay sensitivity due to the low specific activity of radioisotopic labels does *not* often arise, in practice, in the case of competitive assays, whose sensitivity is generally restricted by other factors (Ekins, 1985). The fundamental significance of this conclusion is that, only by the use of labels possessing specific activities higher than those of the commonly used radioisotopes *in assays of non-competitive design*, can current

sensitivity limits be breached. Conversely, use of a higher specific activity label in a *competitive* assay will usually have no significant effect on its sensitivity (assuming experimental errors incurred in reagent manipulation of the magnitude generally encountered in practice).

High specific activity non-isotopic labels

The term 'specific activity' is conventionally applied, in the case of radioisotopic labels, to denote the number of radioactive disintegrations per unit time per unit weight of the isotope or labelled compound. In the present context, use of the term is widened to signify 'detectable events' per unit time per unit weight of labelled material. Thus it can be used to indicate the rate of photon emission by a chemiluminescent or fluorescent label, or the rate of conversion of substrate molecules—by an enzyme label—to molecules of a detectable product. The importance of the concept derives from the fact that 'signal measurement error' (i.e. error in the measurement of the label *per se*) is a contributory factor in limiting assay sensitivity, and may—when other sensitivity-constraining factors are reduced—become dominant. Furthermore, when extending the sensitivities of immunoassay systems beyond their present limits, the numbers of molecules involved are low, and statistical errors incurred in counting individual 'detectable events', and the time required to count them, may assume a particular importance.

Table 1 compares the specific activities of potentially useful labels with that of ^{125}I. All are of relevance in the context of this volume since chemiluminescent and fluorescent labels can be used to label antibodies (or antigens) directly; alternatively, enzyme labels catalysing reactions yielding chemiluminescent signals or fluorescent products can be utilized.

Table 1. Relative specific activities of various isotopic and non-isotopic labels. Note that, though the specific activity of ^{125}I-labelled reagents does not, in practice, significantly limit the sensitivity of competitive assays (see Fig. 4), the lower specific activity of ^{3}H may severely restrict the sensitivity of competitive assays (e.g. of steroid hormones) which rely on the use of this particular radioisotope

Specific Activities	
^{125}I:	1 detectable event/sec/7.5 × 10^6 labelled molecules.
^{3}H:	1 detectable event/sec/5.6 × 10^8 labelled molecules.
Enzymes:	Determined by enzyme 'amplification factor' and detectability of reaction product.
Chemiluminescent labels	1 detectable event/labelled molecule.
Fluorescent labels:	Many detectable events/labelled molecule.

The importance of background in non-competitive immunoassays

A second important factor governing the sensitivity of non-competitive labelled-antibody immunoassays is the 'background' or 'blank' signal emitted in the absence of analyte, since error in the measurement of this signal is clearly a major determinant of the error in measurement of zero dose. Amongst contributors to the background signal are the 'noise' of the measuring instrument itself, 'ambient' signal generators (such as, in 'sandwich' immunoassays, solid 'capture-antibody' supports or, in the case of radioisotopic methods, cosmic ray and other extraneous radiation sources) and 'non-specifically bound' labelled antibody. Minimization of each of these components is essential for maximal sensitivity: mere arithmetic subtraction of background is of absolutely no benefit in this context.

Non-specific binding of antibody is of particular interest, since the magnitude of this contribution is dependent, *inter alia*, on the amount of labelled antibody used in the system, and the duration of its exposure to analyte. Thus increasing the amount of labelled antibody increases the amount of such antibody bound to analyte; however, it may also increase the non-specifically bound moiety to a greater proportional extent, and thus cause a net reduction in sensitivity. This effect underlies the loss in sensitivity at higher antibody concentrations depicted in Fig. 5 (reproduced from Jackson *et al.*, 1983). This phenomenon also underlies the relationship between sensitivity and the affinity constant of the labelled antibody depicted in Fig. 4. The possession by labelled antibody of a high affinity constant implies that a

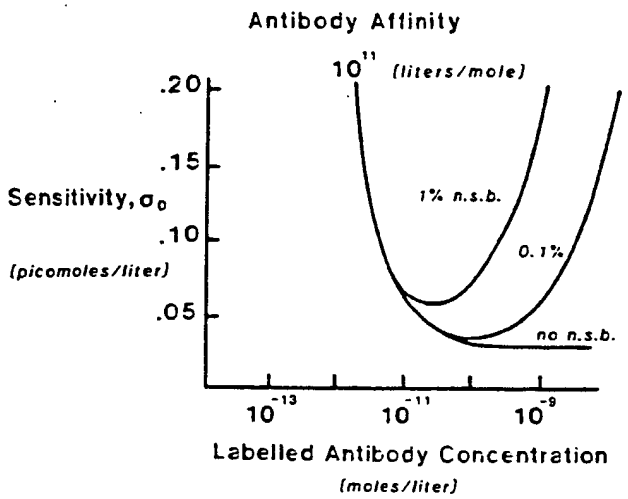

Figure 5. Assay sensitivity (represented by the standard deviation of the zero dose measurement, σ_0), plotted as a function of the concentration of labelled antibody (of affinity 10^{11} L/M) used in the assay, assuming different levels of non-specific binding of labelled antibody. (Note: an irreducible instrument background has been assumed in the computations represented; this limits the ultimate sensitivity attainable, regardless of the concentration of antibody used.)

lower concentration is required to yield the same level of analyte binding, albeit with reduced non-specific binding, thus increasing assay sensitivity

In summary, the high sensitivity of non-competitive labelled antibody methods derives essentially from their permitted use of optimal concentrations of antibody which (provided non-specific binding of labelled antibody is low) are generally considerably greater than in competitive methods, *not* from the fact that the antibody is labelled. Labelled antibody methods generally *fall* in sensitivity as the concentration of antibody is reduced towards zero, ultimately yielding a sensitivity theoretically identical to that of competitive methods (Rodbard and Weiss, 1973). (Paradoxically, early exponents of labelled antibody methods, whilst claiming them to be of higher sensitivity, also concluded that their sensitivity was *increased* by reduction in the amount of labelled antibody used (Woodhead *et al.*, 1971). This incorrect conclusion—based on observation of effects on the slope of the dose-response curve—exemplifies the many fallacies encountered in the immunoassay field stemming from confusion regarding the concept of sensitivity discussed above.) Finally it should be emphasized that maximization of the sensitivity of a non-competitive immunoassay generally implies the selection of reagent concentrations and other experimental conditions such that the [analyte signal/background] ratio (i.e. s/b) is maximized. However, this simple relationship disregards statistical considerations which arise when the numbers of detectable events are very low, and a more appropriate objective may, under these circumstances, be maximization of the ratio s^2/b (Loevinger and Berman, 1951).

Other performance characteristics of competitive and non-competitive immunoassays

Non-competitive designs also display a number of other advantages deriving from the relatively high antibody concentrations on which they generally rely. These include increased reaction speeds (and hence shorter incubation times), decreased vulnerability to certain environmental effects (which cause variations in binding affinity between antibody and analyte), reduced sensitivity-dependence on high antibody binding affinity, etc.

Nevertheless a price has to be paid for these benefits; this includes the greater tendency of a large amount of antibody to bind molecules differing from, but with structural resemblance to, the analyte itself, implying a loss of assay *specificity*. This effect generally necessitates the use, whenever possible, of an 'immunoextraction' procedure using a second 'capture' antibody (usually directed against a different binding site, or 'epitope') as shown in Fig. 3(b). This technique—the 'sandwich' or 'two-site' immunoassay (Wide, 1971)—thus potentially combines the twin virtues of ultra-high sensitivity and specificity (together with short reaction time), features of crucial importance in many diagnostic situations (for example, in the detection of AIDS viral antigens). (Note, however, that the loss of specificity inherent in non-competitive assay designs implies that they are less readily applicable to the measurement of analytes of small molecular size, which cannot be simultaneously bound by two different antibodies directed against different antigenic sites on the molecule. Such analytes are generally more appropriately measured using 'competitive' assay methods.)

Development of ultra-sensitive immunoassay methodologies

The perception that the development of 'ultra-sensitive' immunoassay systems (i.e. systems surpassing conventional RIA methods in sensitivity) depends on (a) reliance on 'excess reagent' or 'non-competitive' assay designs; (b) the use of non-isotopic labels displaying higher specific activities than commonly used radioisotopes; (c) the development of efficient separation systems (ensuring minimization of non-specific antibody binding, and hence of signal 'backgrounds'), and (d) dual or multi-antibody analyte-recognition systems (exemplified by 'sandwich' or two-site assays) to maintain/increase assay specificity, has formed the basis of our own laboratory's immunoassay development since the early to mid-1970s (Ekins, 1978). This led us, *inter alia*, to an immediate recognition (Ekins, 1979, 1980) of the importance of the *in vitro* techniques of monoclonal antibody production pioneered by Köhler and Milstein (1975), which are currently the subject of bitter patent disputes in the USA (Ezzell, 1986, 1987a,b), and which may be expected in Europe.

Meanwhile, of the candidate labels for use in this context, both chemiluminescent and fluorescent labels offer many attractions. The development of stable, highly chemiluminescent, acridinium esters by McCapra and his colleagues (McCapra *et al.*, 1977) has subsequently been exploited by Weeks *et al* (1983, 1984) and, more recently, by several commercial kit manufacturers; other workers have used more conventional chemiluminescent compounds to label immunoassay reagents (see, for example, Kohen *et al.*, 1984, 1985; Barnard *et al.*, 1985). Yet others have relied on enzyme labels to catalyse chemiluminogenic (Whitehead *et al.*, 1983) and fluorogenic (Shalev *et al.*, 1980) reactions as indicated above. Detailed description of these various methodologies is presented by others in this volume and need not be duplicated here.

Common to all the 'ultra-sensitive' immunoassay methodologies relying on such alternative labels is their dependence on a non-competitive, labelled antibody, assay strategy whenever appropriate; however, for the reasons indicated above, *competitive* methods continue to be generally employed for the measurement of analytes of small molecular size (e.g. therapeutic drugs, steroid and thyroid hormones, etc.). Nevertheless, the convenience (from a manufacturing viewpoint, and for other technical reasons) of relying on standard labelling procedures has meant that, even in these cases, labelled antibody techniques are increasingly preferred. Though the commercial kits based on these various labels differ to a minor extent in sensitivity, specificity, convenience, etc., such differences are at least partially attributable to differences in the physicochemical characteristics of the antibodies used in the kits, and to other 'immunological' factors unconnected with the particular nature of the label *per se*.

Despite the obvious attractions of chemiluminescent techniques in an immunoassay context, the use of fluorescent labels combined with sophisticated time-resolution techniques for their detection (a concept arising from discussions with J. F. Tait in 1970) appeared to us (in the mid-1970s) to offer more exciting long-term possibilities for a number of reasons. These naturally included attainment of the enhanced specific activities and high signal to background ratios required for ultra-sensitive immunoassay as indicated above. However, more importantly, fluorescence techniques also appeared to provide a simple route to the development of 'multi-analyte' assay systems of the kind described below.

In pursuance of this strategy, we began collaboration with LKB/Wallac, *ca* 1976–77, in the development of the instrumentation and technology required to develop such methods. Fortunately a group of fluorescent substances generally known as the lanthanide chelates (including, in particular, the chelates of europium, samarium and terbium facilitate such development, possessing prolonged fluorescence decay times (~10–1000 μs), large Stokes shift (~300 nm) and other desirable physical characteristics which permit the construction of relatively cheap instrumentation for their measurement (Marshall *et al.*, 1981; Hemmilä *et al.*, 1983). The fluorescent properties of the lanthanide chelates may be compared with those of a conventional fluorophor such as fluorescein which is characterized by a much smaller Stokes shift (~28 nm), and a fluorescent decay time and emission spectrum which imply that it is less readily distinguished from fluorescent substances present in blood (such as bilirubin) or in plastic sample holders. The unique fluorescence characteristics of the lanthanide chelates thus permit them to be

measured in the presence of a fluorescence background (deriving from extraneous sources) which, in practice, approaches zero. Fig. 6 illustrates the basic concepts involved in pulsed-light, time-resolved, fluorescence measurement, which form the basis of the DELFIA immunoassay system currently marketed by LKB/Wallac.

Though it is inappropriate to pursue this subject in greater detail, attention should also be drawn to the possibilities offered by phase-resolved fluorimetry. This permits separate identification of fluorophores differing in fluorescence lifetime by their exposure to a sinusoidally modulated exciting light source, and observation of their demodulated, phase-shifted, light emission (McGown and Bright, 1984). This technique offers the possibility both of the development of homogeneous assays (relying on a difference in fluorescence decay time of bound and free forms of the fluorescent-labelled molecule), and of discriminating between two labelled antibodies in the context of multi-analyte 'ratiometric' immunoassay as discussed below.

'AMBIENT ANALYTE' IMMUNOASSAY

Before proceeding to a discussion of the development of multi-analyte assays, another important concept, termed 'ambient analyte immunoassay' (Ekins, 1983b), must first be examined. This term is intended to describe a type of immunoassay system which, unlike unconventional methods, measures the analyte *concentration* in the medium to which an antibody is exposed, being essentially independent both of sample volume, and of the amount of antibody present. This concept is illustrated in Fig. 7, and relies on the physicochemically-based proposition that, when a 'vanishingly small' amount of antibody (preferably, but not essentially, coupled to a solid support) is exposed to an analyte-containing medium, the resulting (fractional) occupancy of antibody binding sites solely reflects the ambient analyte concentration. Clearly the binding by antibody of analyte results in a depletion of the amount of analyte in the surrounding medium, but provided the proportion so bound is small (i.e. less than, for example, 1% of the total), such disturbance can be ignored. (This effect is closely analogous to that caused by the introduction of a thermometer into a medium possessing a much larger thermal capacity; the temperature disturbance caused by the thermometer itself is negligible and can, in these circumstances, be disregarded.)

The principles of ambient analyte assay derive from the recognition that *all* immunoassays essentially depend upon measurement of the 'fractional occupancy' by analyte of antibody binding sites following reaction of analyte with antibody as discussed above (Figs 3. (a) and (b)). The fractional occupancy of ('monospecific' or 'monoclonal') antibody binding sites in the presence of varying analyte concentrations, plot-

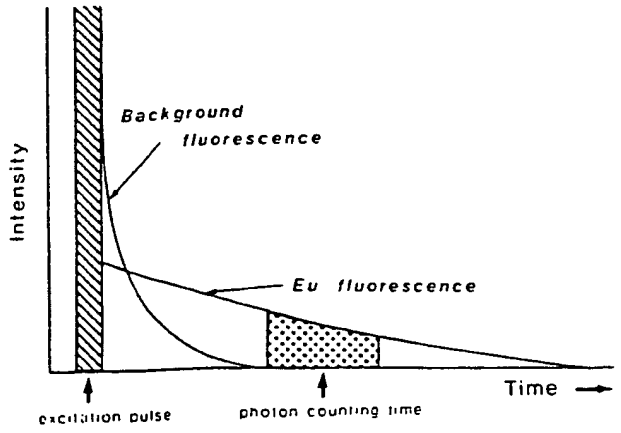

Figure 6. Basic principles of pulse-light, time resolved fluorescence. Fluorescence emitted by the fluorophor (typically a europium chelate) is distinguished from background fluorescence, which decays more rapidly.

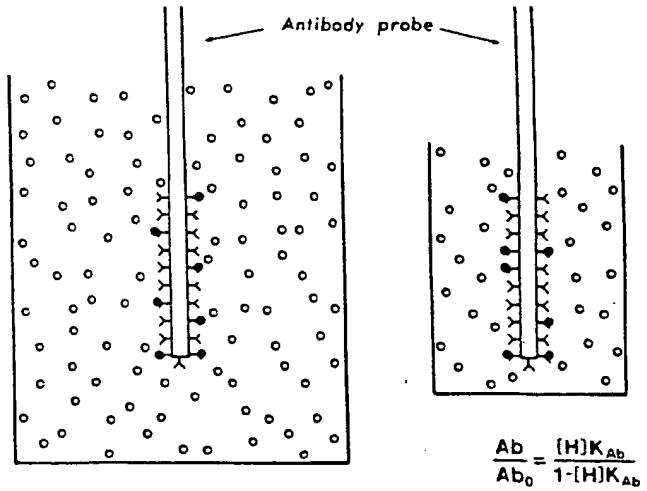

Figure 7. Basic principle of 'ambient analyte' immunoassay (AAI). The fractional occupancy (F) of a vanishingly small amount of antibody (of affinity K) is determined by the analyte concentration in the medium ([An]).

ted against antibody concentration, is portrayed in Fig. 8. The fraction of analyte bound is also plotted in this figure. (Note: for the sake of generality, all concentrations in this figure are expressed in terms of $1/K$, where K is the affinity constant of the antibody. For example, if $K = 10^{11}$ L/M, a concentration of $0.1 \times 1/K$ represents 0.1×10^{-11} M/L, or $0.1 \times 10^{-11} \times 10^{-3} \times 6.02 \times 10^{23} = 6.02 \times 10^{8}$ molecules/ml.)

It should be particularly noted that, at antibody concentrations of less than *ca* $0.01 \times 1/K$ antibody fractional occupancy is essentially dependent solely on the analyte concentration in the medium, and is independent of variations in antibody concentration. This reflects the fact that this concentration of antibody binds less than approximately 1% of the analyte in the medium, irrespective of its concentration. This implies, for example, that the introduction of 10, 100, or 1000 antibody molecules into a medium containing billions of analyte molecules will result, in each case, in virtually identical fractional antibody binding-site occupancy, the upper limit of antibody concentration being determined by the antibody affinity constant. (An antibody concentration of $0.01 \times 1/K$ is a hundred-fold less than that ($1 \times 1/K$) necessary to bind 50% of a 'trace' amount of analyte (see Fig. 8), claimed by Berson and Yalow (1973) as maximizing assay 'sensitivity' (i.e. the slope of the dose–response curve when expressed in terms of bound/free labelled analyte). This false conclusion has subsequently become incorporated into the mythology of radioimmunoassay design which, regrettably, a majority of kit manufacturers continue to accept.)

The ambient analyte assay concept was originally exploited in the original development of what has come to be known as 'two-step' free hormone immunoassay (Ekins *et al.*, 1980), but it is clear that it is of far wider application, and can, in particular, be utilized in the construction of immunosensors and immunoprobes. One such example is a probe for the measurement of salivary steroids that is currently being developed in our laboratory. Comprising a small antibody-coated plastic 'dipstick' comparable in size and shape to a clinical thermometer, this device is intended to permit the measurement of salivary steroid levels without requiring the collection of saliva. However, the concept also underlies our approach to multi-analyte immunoassay, also under development in our laboratory.

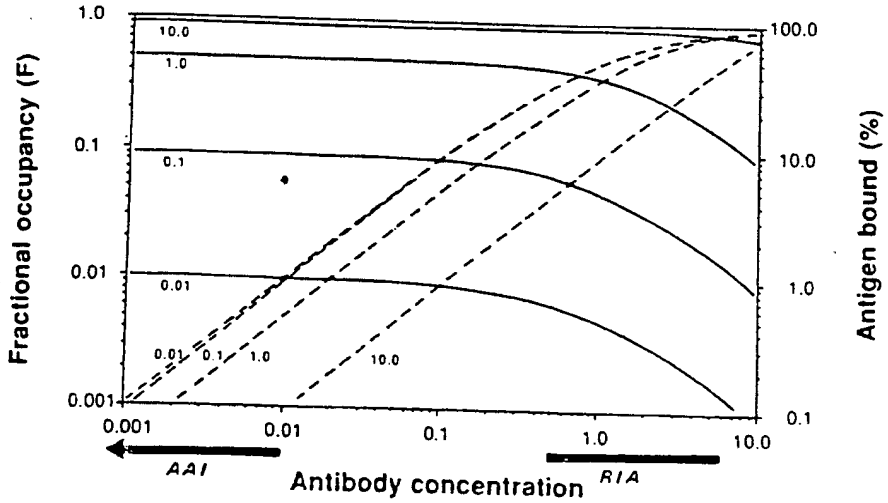

Figure 8. Fractional antibody binding-site occupancy (F) plotted as a function of antibody binding-site concentration for different values of analyte (antigen) concentration [An]. The percentage binding of analyte to antibody (b) is also shown. All concentrations are expressed in units of $1/K$. Note that for antibody concentrations of less than $0.01/K$ (approximately), percentage binding of analyte is <1%, and fractional binding-site occupancy is essentially unaffected by variations in antibody concentration extending over several orders of magnitude, being governed solely by [An]. Note that radioimmunoassays and other 'competitive' immunoassays are commonly designed using antibody concentrations approximately $0.5/K$–$1/K$ or above implying $b_0 > 30\%$), in accordance with the precepts of Berson and Yalow (e.g. Berson and Yalow, 1973).

MULTI-ANALYTE 'RATIOMETRIC' IMMUNOASSAY SYSTEMS

The concepts relating to ambient analyte immunoassay and assay sensitivity outlined above are both exploited in our present development of a random access, multi-analyte, immunoassay technology capable of measuring, in the same small sample, virtually any number of individual analytes from selected analyte 'menus' (e.g. a hormone menu, viral antigen menu, an allergen menu, etc.). Many examples of a need to measure a multiplicity of different analytes in the same sample exist in medical diagnosis, for example, in the routine diagnosis of thyroid disease, where it is frequently necessary to measure a number of different hormones and thyroid-related proteins. At present, clinicians frequently experience difficulty in deciding on the best sequence of tests to arrive at a correct diagnosis. Such problems would be overcome were all relevant analytes measurable at a cost comparable to the cost of measurement of a single substance. Our own immediate objective is the development of a technology permitting the measurement of complete 'hormone profiles' using a single small blood sample. However, the need for 'multi-analyte', or 'random access' measurement is not confined to medical diagnosis: it also arises, for example, in the pharmaceutical industry (where there exists a requirement to ensure the purity of protein drugs synthesized by recombinant DNA techniques), in the food industry and elsewhere. Though still at an early stage, our approach to the achievement of this objective can be briefly indicated.

Multi-analyte assay: general principles

As discussed above, the notion of ambient analyte assay simultaneously introduces two extremely important and novel concepts: (a) that an estimate of analyte concentration can be based upon the use of an infinitesimal amount of 'sampling' antibody, and (b) that such an estimate derives from a direct measurement of fractional antibody occupancy by analyte, irrespective of the exact amount of antibody used. It should be emphasized that the latter proposition is valid only in the context of ambient analyte assay, and is *not* true in current conventional immunoassay systems (in which fractional antibody occupancy depends both upon the amount of antibody in the system, and sample volume—see Fig. 8). In short, exposure of a small number of antibody molecules (in the form, for example, of a 'microspot' located on a solid support) to an analyte-containing fluid results in occupancy of antibody binding sites in the microspot reflecting the analyte concentration in the medium. Following such exposure, the antibody-bearing probe may be removed and exposed to a 'developing' solution containing a high concentration of an appropriate second antibody directed against either a second epitope on the analyte molecule if this is large (i.e. the occupied site), or against unoccupied antibody binding sites in the case of small analyte molecules (see Fig. 3(b)). (Note: an antibody simulating antigen, and reacting with unoccupied binding sites, is described as a 'mirror-image anti-idiotypic antibody'; the use of such an antibody instead of labelled antigen is convenient but not essential, and is suggested here merely to simplify illustration of the basic concepts involved.)

Subsequently, an estimate of binding-site occupancy of the 'sampling' (solid phase) antibody located in the microspot may be derived by measurement of the ratio of signals emitted by the two antibodies forming the dual-antibody 'couplets'. This can be conveniently achieved by labelling the 'sampling' and 'developing' antibodies with different labels, for example, a pair of radioactive, enzyme or chemiluminescent markers. Fluorescent labels are nevertheless particularly useful in this context because, by the use of optical scanning techniques, they permit arrays of different antibody 'microspots' distributed over a surface, each directed against a different analyte, to be individually examined, thus enabling multiple assays to be simultaneously carried out on the same small sample. Fig. 9 illustrates these basic ideas, and Fig. 10 such an array.

Microspot immunoassay sensitivity: theoretical considerations

The notion that it is, in principle, possible to measure an analyte concentration using a microspot of antibody comprising a number of antibody molecules in the range *ca* 10^1–10^6 is likely, at first sight, to appear surprising, and may, indeed, provoke scepticism regarding the assay sensitivities potentially attainable using this approach. Clearly a number of factors, such as the sensitivity

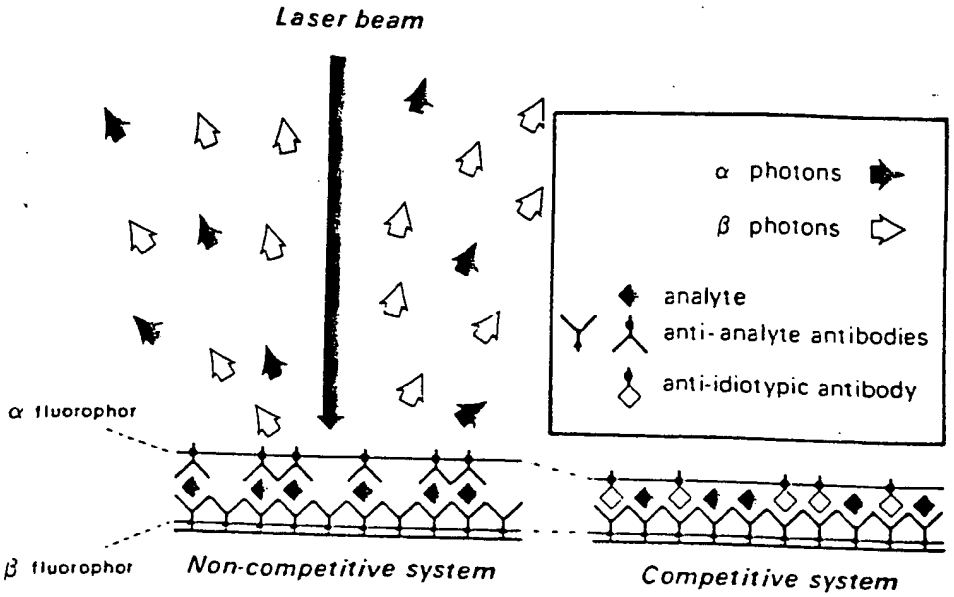

Figure 9. Basic principle of dual-label, ambient-analyte, immunoassay relying on fluorescent labelled antibodies. The ratio of α and β fluorescent photons emitted reflects the value of F (see Figs 5 and 6) and is solely dependent on the analyte concentration to which the probe has been exposed. It is unaffected by the amount or distribution of antibody coated (as a monomolecular layer) on the probe surface.

of the signal measuring equipment, the density of antibody molecules on the surface of the solid support, etc., are likely to play a part in determining final assay sensitivity. Such factors are, in turn, dependent on the efficiency with which the particular labels used can be detected, the adsorption properties of antibody supports, etc. Though these are obviously variable, reasonable estimates can be made of the order of sensitivities likely to be achieved on the basis of some simple theoretical calculations. To clarify the following discussion, it is assumed that 'sensing' antibody can be uniformly and consistently coated on a solid matrix at a standard density, implying that only the 'developing' antibody need be labelled and measured in order to ascertain fractional occupancy of sensing antibody binding sites.

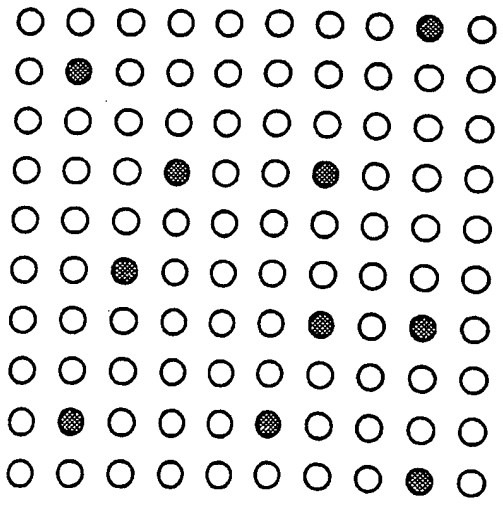

Figure 10. 'Multi-analyte' antibody array. Each antibody 'microspot' represents a 'vanishingly small' amount of antibody directed against an individual analyte.

Fig. 11 illustrates the surface of an antibody microspot, of surface area $A(\mu m^2)$, and (uniformly) coated with antibody of affinity K(L/M) in a monomolecular layer of density D(molecules/μm^2). Let us assume that the spot is exposed to an analyte-containing medium of volume v(ml), and containing an analyte concentration C molecules/ml. The molecular concentration of antibody in the system is thus given by AD/v. (Note: the fact that antibody is situated on the surface of a solid support, and not evenly distributed throughout the medium, does not affect the extent of analyte binding at thermodynamic equilibrium, assuming that antibody binding sites are not impeded in their reactions and have not been damaged during the coating process.)

Meanwhile, fractional occupancy (F) of antibody binding sites by analyte (at equilibrium) is

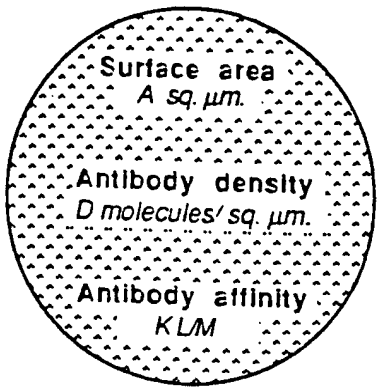

Figure 11. Microspot ambient-analyte immunoassay. The microspot shown is assumed to be uniformly coated with antibody, though if the dual-labelled antibody 'ratiometric' approach shown in Fig. 9 is adopted, uniform coating is not essential. The minimum fluid volume for ambient analyte assay conditions to prevail (enabling adoption of the ratiometric approach) is shown. Minimum test sample volume (M/S): $A \times D \times K \times 10^5/N$

given by the equation:

$$F^2 - F(1/q + p/q + 1) + p/q = 0 \qquad (1)$$

where p = analyte concentration, q = antibody concentration (both expressed in units of $1/K$).

Thus, for antibody binding site concentrations $\rightarrow 0$ (i.e. $q < 0.01$), $F \approx p/(1 + p)$; (see Fig. 8).

Likewise, the fraction of analyte bound by antibody (f) at equilibrium is given by the equation:

$$f^2 - f(1/p + q/p + 1) + q/p = 0 \qquad (2)$$

Thus, for analyte concentration $\rightarrow 0$ (i.e. $p < 0.01$), $f \approx q/(1 + q)$; (see Fig. 8). Furthermore, when $q < 0.01$, and when $p \geq 0$, $f < 0.01$.

Expressed in units of $1/K$; the concentration (q) in the assay of 'sensing' antibody situated on the microspot is given by $DAK/(\nu \times 6 \times 10^{20})$, (since Avogadro's constant, expressed as the number of molecules/mmol, is 6×10^{20} (approximately)). The fraction of an analyte concentration $\rightarrow 0$ which will be bound to the spot is therefore $DAK/(\nu \times 6 \times 10^{20} + DAK)$, implying that the number of analyte molecules bound to the spot is given by $\nu CDAK/(\nu \times 6 \times 10^{20} + DAK)$.

Case 1: sandwich (two-site) assay. Following incubation of sample with antibody, we assume the sample is removed, and the microspot then exposed to a volume V(ml) of a solution of a second, labelled, 'developing' antibody of affinity K^* (L/M) at a concentration given by Q (expressed in units of $1/K^*$).

The fraction of analyte bound by labelled antibody (F^*) at equilibrium is given by the equation:

$$F^{*2} - F^*(1/P + Q/P + 1) + Q/P = 0 \qquad (3)$$

where P represents the analyte concentration in the developing-antibody solution, expressed in units of $1/K^*$, i.e. $\nu CDAKK^*/[(\nu \times 6 \times 10^{20} + DAK)V \times 6 \times 10^{20}]$.

Assuming $P < 0.01$, $F^* \approx Q/(1 + Q)$. (For example, if $Q = 1$, the fraction of analyte molecules bound by labelled antibody = 0.5 approximately). Thus, since the number of analyte molecules bound to the spot is given by $\nu CDAK/(\nu \times 6 \times 10^{20} + DAK)$, the number of analyte molecules labelled by the second, developing, antibody is given by $\nu CDAKQ/[(\nu \times 6 \times 10^{20} + DAK)(1 + Q)]$, and the surface density of such molecules is given by $\nu CDKQ/[(\nu \times 6 \times 10^{20} + DAK)(1 + Q)]$. Moreover, assuming that $DAK \ll \nu \times 6 \times 10^{20}$ (i.e. that the amount of antibody in the system is such that 'ambient assay' conditions prevail, then the surface density (D^*) of developing-antibody molecules = $CDKQ/[(6 \times 10^{20})(1 + Q)]$ approximately. It should be noted that D^* is independent of both ν and V, also that the ratio $D^*/D = C \times KQ/[(6 \times 10^{20})(1 + Q)] = C \times$ constant.

If the minimum detectable surface density of developing-antibody molecules (i.e. σ_{D0}^*, the standard deviation of the measurement of D^* when $C = 0$) is given by D^*_{min} (molecules/μm^2) and C_{min} represents the minimum detectable analyte concentration in the test sample, then,

disregarding non-specific binding of developing antibody within the microspot area,

$$C_{min} = D^*_{min} \times [(6 \times 10^{20})(1 + Q)]/DKQ \quad (4)$$

For example, if $Q = 1$, $D = 10^5$ molecules/μm^2, $K = 10^{11}$ L/M and $D^*_{min} = 20$ molecules/μm^2, then $C_{min} = 2.4 \times 10^6$ molecules/ml = 10^{-15} M/L. It should be noted, in this example, the fractional occupancy of the sensing antibody binding sites by the minimum detectable analyte concentration is 0.04%.

Case 2: anti-idiotypic antibody ('competitive') assay. In this case, we assume that, following removal of the sample, the microspot is exposed to a volume V(ml) of a solution of (for example) a second, labelled, anti-idiotypic antibody reacting with *unoccupied* sites on the sensing antibody. Using similar reasoning as above, we may likewise assume that the fraction of such sites which become occupied by the anti-idiotypic 'developing' antibody is given by $Q/(1 + Q)$, where Q is the developing-antibody concentration. However, the minimum detectable surface density of anti-idiotypic antibody is not, in a competitive design, the critical determinant of assay sensitivity; this parameter is essentially governed by the precision of the density measurement.

From Eq. (1), the fraction of sites *unoccupied* by analyte = $1/(1 + p)$, and the fraction occupied by anti-idiotypic antibody = $Q/(1 + p)(1 + Q)$. Thus, if the CV in the measurement of anti-idiotypic antibody is ε, the standard deviation is $\varepsilon Q/(1 + p)(1 + Q)$. This term also represents the SD in the estimate of the fraction of sites *occupied* by analyte. Since the total number of antibody binding sites in the spot is DA, the SD in the estimate of occupied sites as $p \rightarrow 0$ (i.e. σD^*_0) approximates $\varepsilon DAQ/(1 + Q)$; the SD in the occupied site surface-density estimate is thus $\varepsilon DQ/(1 + Q)$. But the SD in the measurement of fractional binding-site occupancy when $p \rightarrow 0$ defines D_{min}, and hence the minimum detectable analyte concentration in the test sample as indicated in Eq (4).

Thus

$$C_{min} = D_{min} \times [(6 \times 10^{20})(1 + Q)]/DKQ \quad (5)$$

$$= \varepsilon DQ/(1 + Q) \pm [(6 \times 10^{20})(1 + Q)]\, DKQ \quad (6)$$

$$= \varepsilon/K \times (6 \times 10^{20}) \quad (7)$$

For example, if values of $Q = 1$, $D = 10^5$ molecules/μm^2, and $K = 10^{11}$ L/M are assumed as in the non-competitive example considered above, and the CV in the measurement of anti-idiotypic antibody density in the microspot is 1% (i.e. $\varepsilon = 0.01$), then $D_{min} = 500$ molecules/μm^2, and $C_{min} = 6 \times 10^7$ molecules/ml = 10^{-13} M/L. Fractional occupancy of the sensing antibody binding sites by the minimum detectable analyte concentration is, in this example, 1%. It should be noted that the sensitivity limit of ε/K (expressed in molar terms) is identical to that previously established for conventional 'competitive' assays (Ekins and Newman, 1970), and which underlies the predictions represented in Fig. 4.

Such considerations appear to suggest (a) that microspot assay sensitivities superior to those obtainable by conventional radioisotopically based immunoassays are achievable, and (b) that sensitivities yielded by non-competitive microspot assays are likely to be considerably greater than those of corresponding competitive microspot assays. It must be emphasized, however, that, though such predictions are likely to prove correct, assumptions regarding the performance of the labels and signal-measuring instrument used are incorporated in the simple theoretical analysis discussed above. Such factors are clearly of importance in determining overall microspot immunoassay performance.

Practical implementation

The concepts discussed above are clearly exploitable using a variety of antibody labels, including chemiluminescent labels; however, our preliminary studies have been based on the use of conventional fluorophores, since the technology of simultaneous measurement of dual fluorescence from small areas is already well established. Because this volume centres on chemiluminescence, we shall provide only a brief indication of our initial experimental work in this area, which is currently based on the use of commercially available confocal microscopes.

Instrumentation: the laser scanning confocal microscope. In laser scanning confocal fluoresc-

ence microscopy, a small area of the specimen is illuminated by a focused laser beam; the fluorescence photons emanating solely from this area are, in turn, focused onto a photon detector. Both the intensity of illumination and the efficiency of light collection diminish rapidly with distance from the focal plane (Fig. 12). At the 'confocal' point, the projection of the illumination pinhole and the back-projection of the detector pinhole coincide. Such systems contrast with conventional epi-fluorescence methods, where the specimen is exposed to an essentially uniform flux of illumination (White *et al.*, 1987).

Sensitivity of current instruments. Typically, fluorescence photons emanating from the laser-illuminated area are detected by a low dark-current photomultiplier. Electrons spontaneously emitted by the photomultiplier photocathode contribute to the background signal of the instrument, and must, for highest sensitivity, be minimized. Fortunately the overall design of such instruments permits the photomultiplier photocathode to be of very small area, so that this particular source of background noise is not only small, but can be expected to reduce in relative importance with future improvement in photomultiplier design. Meanwhile current instruments already display very high sensitivity of detection of fluorescent signals. For example, the confocal microscope manufactured by Zeiss is claimed to display a lower detection limit for fluorescein of about ten molecules/μm^2 (Ploem, 1986). Most commercially available FITC-labelled IgG attains a fluorophore/protein molar ratio of ~4; thus the detection limit (D^*_{min}) of the Zeiss microscope is ~2–3 FITC-labelled IgG molecules/μm^2. This implies an analyte-concentration detection limit of ~2.4×10^5 molecules/ml for a two-site assay, assuming the same parameter values as used in the examples discussed above, or 2.4×10^4 molecules/ml using a 'sensing' antibody of affinity 10^{12} L/M.

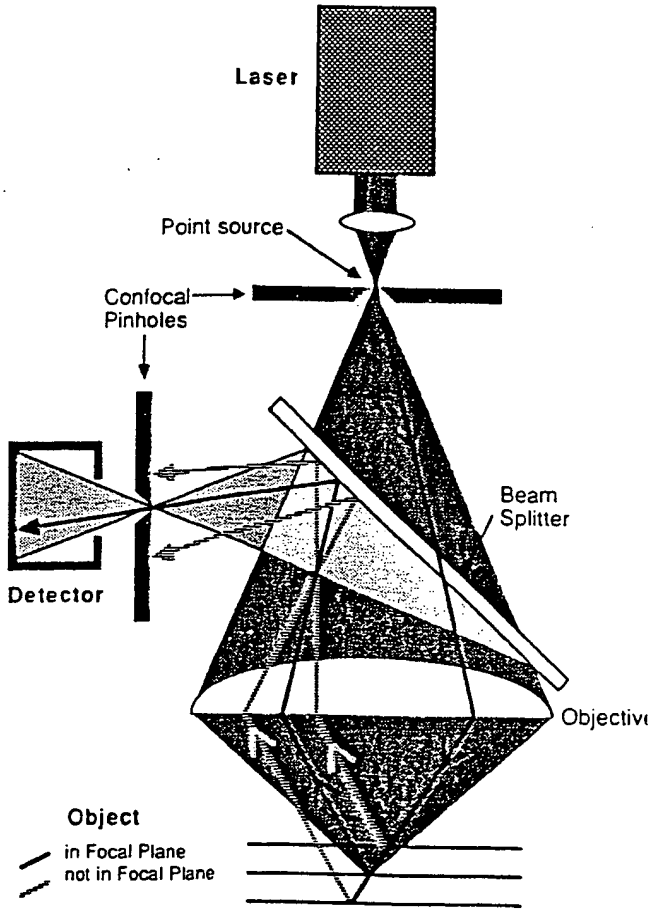

Figure 12. Principle of the confocal microscope. Illuminating light is focused at a point in the focal plane. Reflected light from this point is focused onto a detector. A complete two-dimensional image of structures within the focal plane is obtained by scanning the selected area of interest, and may be stored in a microcomputer for video display

Another comparable instrument is the Bio-Rad/Lasersharp laser scanning confocal microscope, which we are currently using in the development of 'ratiometric' multi-analyte assay methodology in accordance with the principles outlined above (see Fig. 13). The argon laser in this system possesses two excitation lines at 488 and 514 nm. It is thus particularly efficient for the excitation of blue/green emitting fluorophores such as FITC (which displays an excitation maximum at 492 nm). However, it is considerably less efficient in the excitation of red-emitting fluorophores such as Texas red (excitation maximum 596 nm). However, the ratiometric immunoassay principle permits considerable variation in detection efficiencies of the two labels relied on since, *inter alia*, the specific activities of the two labelled antibody species forming the antibody couplets can be chosen to yield optimal signal ratios in the region of unity. Thus inefficiency of the argon laser in exciting red emitting fluorophores is not necessarily a major handicap in the present context.

Though the current Lasersharp instrument relies on a conventional microscope rather than a purpose-designed optical system (and appears to

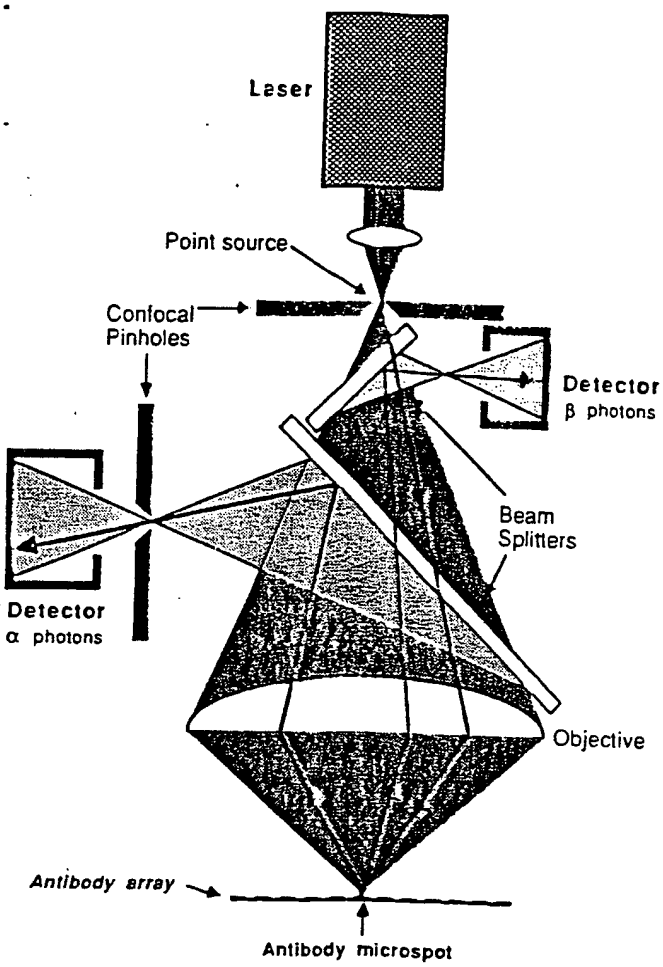

Figure 13. Dual-channel confocal fluorescence microscope permitting simultaneous measurement of the fluorescence signals from two fluorophors situated at the focal point. By scanning the antibody array, the ratio of signals from each antibody microspot may be determined

be less sensitive), it permits quantification of fluorescence signals generated from microspots of selected area. Initial studies have revealed that, under conditions that are not necessarily optimal, the instrument is capable of detecting approximately twenty-five FITC-labelled IgG molecules/μm^2, scanning an area of $\sim 50 \mu m^2$ (Fig. 14). It must be stressed that neither of these confocal microscopes are designed specifically for routine ratiometric multi-analyte immunoassay use, and it can be anticipated that future instruments constructed specifically for this purpose are likely to prove both cheaper and more sensitive.

Other instruments. The MPM 200 Microscope Photometer manufactured by Zeiss of West Germany is anticipated to become available shortly. This photometer is claimed to be highly versatile: it can be used in transmission and reflection modes, and as a highly sensitive fluorimeter. The measuring field can be varied in shape and size for optimum adjustment to the specimen structure. More generally, the technology of sensitive light measurement is improving rapidly in response to needs in astronomy, the space program etc., such technology clearly being readily exploitable in a multi-analyte immunoassay context using light-generating labels in accordance with the broad principles presented here.

Solid antibody supports. On the basis of the theoretical considerations discussed above, it is evident that solid antibody supports for multi-analyte immunoassay use should display a capacity to adsorb a high surface density of antibody combined with low intrinsic signal-generating properties (for example, low intrinsic fluorescence), thus minimizing background. We have examined a number of materials, including polypropylene, Teflon, cellulose and nitrocellulose membranes and microtitre plates (clear polystyrene plates from Nunc; black, white and clear polystyrene plates from Dynatech with these criteria in mind. White Dynatech Microfluor microtitre plates, formulated specially for the detection of low fluorescence signals, yield high signal-to-noise ratios and have therefore been provisionally used in our developmental studies.

Surface density of antibody coating. Preliminary experiments using Microfluor plates have revealed that it is possible to coat them with antibody at a surface density of at least 5×10^4 IgG molecules/μm^2 (Fig. 15). Moreover nearly all antibody molecules so deposited appear to retain immunological activity (Fig. 16).

Verification of the 'ratiometric' imunoassay concept. Our primary intention, in initial studies, has been establishment of the basic conditions which, using a particular instrument, can be anticipated on theoretical grounds to yield high assay sensitivity. Though the setting up of individual microspot immunoassays has thus appeared to us to be of secondary importance during the initial stages of our studies, we have nevertheless

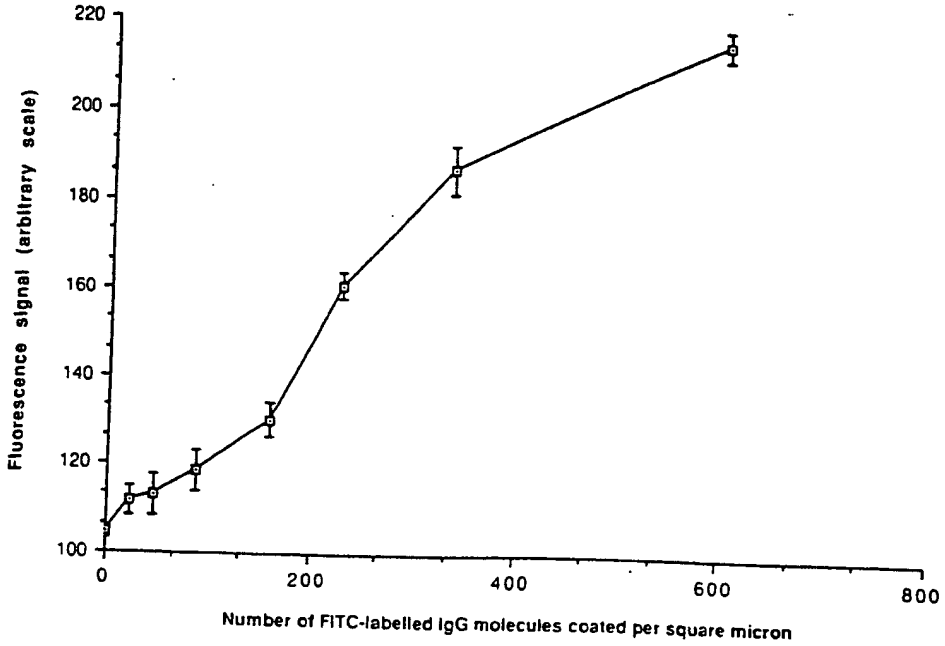

Figure 14. Fluorescence signal (arbitrary units), measured using the Bio-Rad/Lasersharp scanning confocal microscope, plotted as a function of the density of fluorescein-labelled IgG molecules (number of molecules/μm^2) deposited on Dynatech Microfluor white microtitre plates

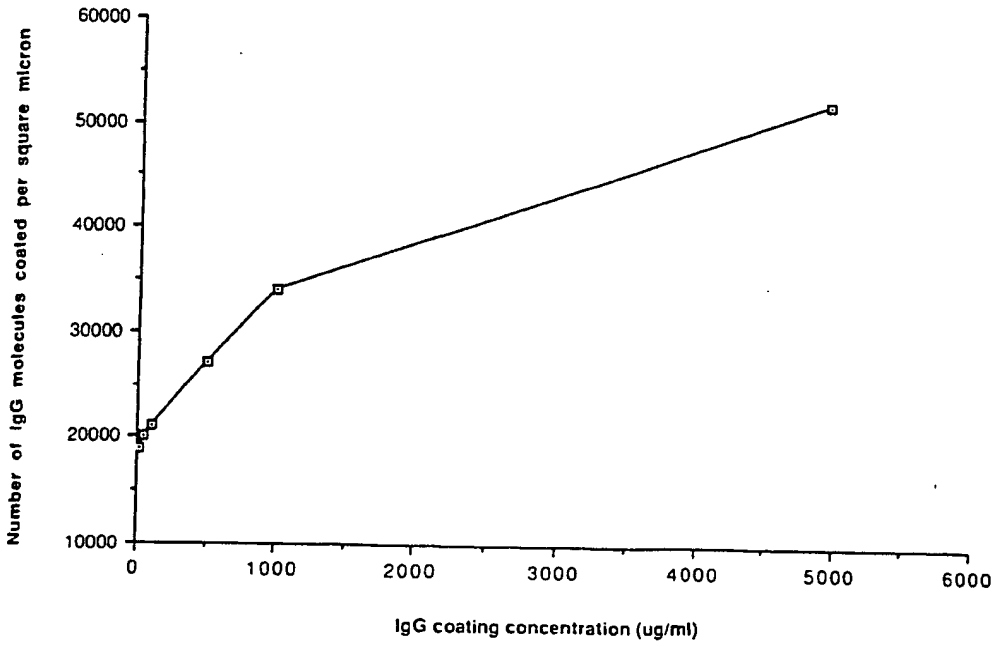

Figure 15. Surface density of IgG molecules (number of molecules/μm^2) deposited on Dynatech Microfluor white plates plotted as a function of IgG concentration ($\mu g/ml$) in the coating solution

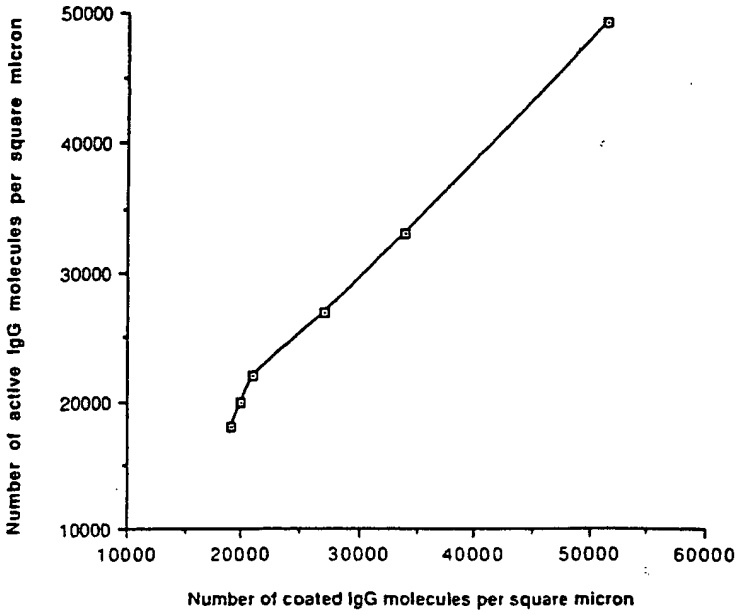

Figure 16. Surface density of immunoreactive IgG molecules (number of molecules/μm^2) plotted as a function of the total surface density of IgG (number of molecules/μm^2) on Dynatech Microfluor white microtitre plates

thought it useful to confirm the validity of our general concepts by comparing the performance of certain assays when constructed in microspot format and when conventionally designed. For example, we have compared a dual-labelled tumour necrosis factor (TNF) ratiometric assay system using Texas red and FITC-labelled antibodies with an optimized IRMA system using identical antibodies but with the second antibody ^{125}I-labelled. Although unoptimized, the ratiometric microspot assay yielded formal sensitivity values closely approaching that of the conventional, optimized, IRMA. Although verifying the general concepts underlying ratiometric microspot immunoassay methodology, further work is required to achieve the considerably greater sensitivity that theory predicts as achievable using optimized reagent concentrations and improved instrumentation.

CONCLUSION

As indicated above, differentiation of the fluorescent signals yielded by two fluorophores can be readily achieved solely on the basis of wavelength differences, and this approach has been relied on entirely in our preliminary studies. However, other physical techniques exploiting differences in decay time of two or more fluorescence emissions (using, for example, a pulsed or sinusoidally modulated laser source, and time– or phase-resolving detectors) are available, and can be expected both to further reduce background and to improve signal resolution, thus increasing assay sensitivity and precision. These considerations aside, the basic technology involved closely resembles that employed in domestic compact disk recorders and other similar data-storage devices, the obvious difference being that light emitted from each of the discrete zones forming the antibody-array is fluorescent rather than reflected, and yields chemical rather than physical information. Indeed, our preliminary studies suggest that highly sensitive immunoassays using antibody microspots of surface area approximating 50 μm^2 are achievable, implying that some 2,000,000 different immunoassays could, in principle, be accommodated on a surface area of 1 cm^2. Though non-specific binding of a multiplicity of developing antibodies would probably prohibit the use of antibody arrays of this order, it is evident that the technology is capable of encompassing analyte numbers of the kind likely to be useful in practice.

The development of multi-analyte assay systems of this kind can be anticipated to bring about

fundamental changes in medical diagnosis and many other biologically related areas. Systems capable of measuring every hormone and other endocrinologically related substance within a single small sample of blood are within technological reach, providing data which, when analysed with the aid of computer-based 'expert' pattern-recognition systems, are likely to reveal endocrine deficiences only dimly perceived using current 'single-analyte' diagnostic procedures. Such systems also provide a means to the development of a 'random access' immunoassay methodology, permitting the selection of any desired test or combination of tests from an extensive analyte menu. Clearly the accommodation of a wide range of individual immunoassays on a small immunoprobe (comparable in its overall physical dimensions with a few drops of blood) is likely to totally transform the logistics of immunodiagnostic testing, and genuinely represents, in our view, 'next generation' immunoassay methodology.

Acknowledgement

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